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(71) Applicant: PICKER INTERNATIONAL, INC.
Highland Heights Ohio 44143 (US)

(72) Inventors:
- Petropoulos, Labros S.
  Solon, Ohio 44139 (US)
- Mastandrea, Nicholas J.
  Euclid, Ohio 44117 (US)
- Richard, Mark A.
  South Euclid, Ohio 44121 (US)

(74) Representative: Waters, Jeffrey
Marconi Intellectual Property
Waterhouse Lane
Chelmsford Essex CM1 2QX (GB)

(54) MRI gradient coil assembly

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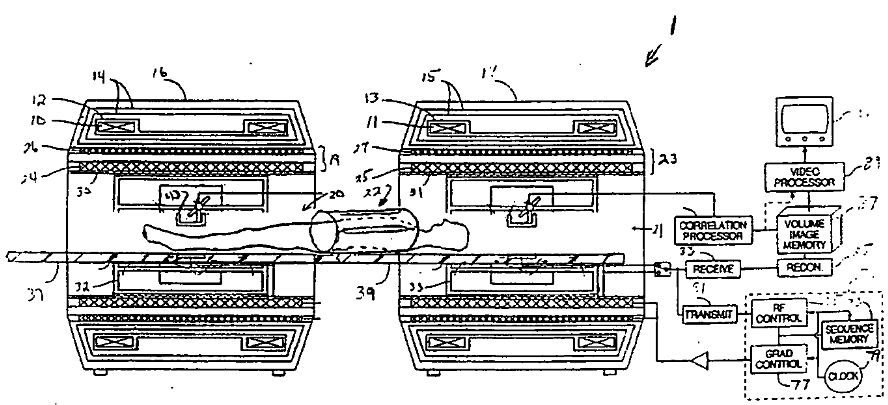

FIG. 1

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EUROPEAN SEARCH REPORT

Application Number
EP 98 31 0334

DOCUMENTS CONSIDERED TO BE RELEVANT

Category	Citation of document with indication, where appropriate, of relevant passages	Relevant to claim	CLASSIFICATION OF THE APPLICATION (Int.Cl.6)
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The present search report has been drawn up for all claims

Place of search	Date of completion of the search	Examiner
MUNICH	12 February 2001	Skalla, J

CATEGORY OF CITED DOCUMENTS

X : particularly relevant if taken alone
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A : technological background
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P : intermediate document

T : theory or principle underlying the invention
E : earlier patent document, but published on, or after the filing date
D : document cited in the application
L : document cited for other reasons

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ANNEX TO THE EUROPEAN SEARCH REPORT ON EUROPEAN PATENT APPLICATION NO. EP 98 31 0334

This annex lists the patent family members relating to the patent documents cited in the above-mentioned European search report. The members are as contained in the European Patent Office EDP file on
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12-02-2001

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WO 9735214 A	25-09-1997	US 5729141 A	17-03-1998

For more details about this annex : see Official Journal of the European Patent Office, No. 12/82

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(74) Representative: **Waters, Jeffrey**
GEC PATENT DEPARTMENT,
Waterhouse Lane
Chelmsford, Essex CM1 2QX (GB)

(54) **MRI gradient coil assembly**

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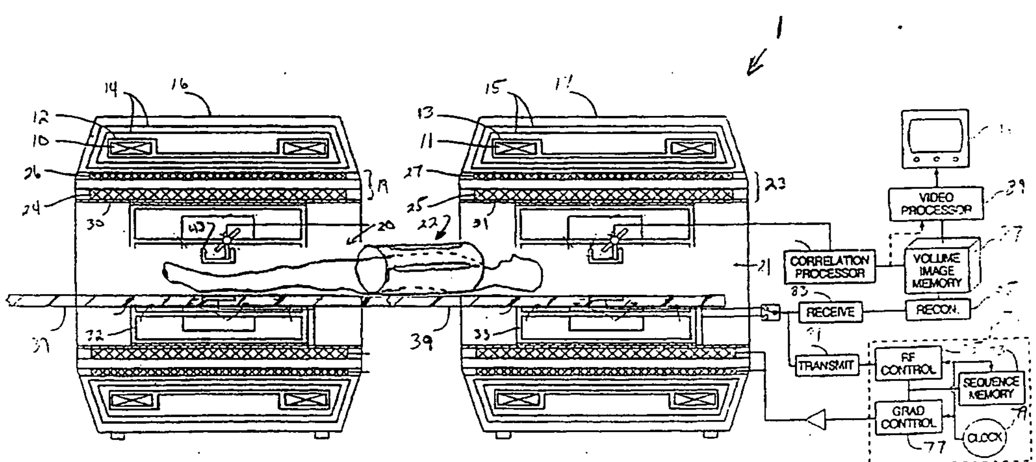

FIG. 1

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Description

[0001] The present invention relates to magnetic resonance imaging. It finds particular application in conjunction with horizontal open magnet imaging equipment of the type used to perform interventional procedures on the torso and head of human subjects and will be described with particular reference thereto. It is to be appreciated, however, that the present invention also finds application in conjunction with imaging and interventional procedures performed on other human body parts and further, in conjunction with the imaging or analysis of non-human and inanimate subjects.

[0002] In magnetic resonance imaging, dipoles are selectively aligned with a primary magnetic field. Radio frequency excitation pulses are applied to stimulate resonance in the aligned dipoles and radio frequency magnetic resonance signals are collected from the resonating dipoles. Gradient magnetic field pulses are applied to encode spatial position. When imaging the human upper torso, which includes the heart, lungs, and other moving tissue, high speed image acquisition is essential.

[0003] To promote high speed image acquisition and high resolution, high strength magnetic field gradients with high slew rates are advantageous. That is, gradients of large magnitude that can be switched on and off very quickly are desirable for improved data acquisition time and resolution. However, gradient strength varies inversely as the radius squared of the gradient coil. Stored energy, a critical factor for slew rate. varies with the fifth power of the radius of the gradient coil. Thus, for upper torso imaging using large diameter coils, the width of the patient's shoulders has been a limiting factor in prior systems. Typically, a whole body gradient magnetic field coil is about 65 cm in diameter. To improve the magnetic field gradient characteristics in the upper torso while minimizing the adverse effects of large diameter sizes, elliptical gradient coils and planar gradient coils have been utilized with some success.

[0004] To improve data acquisition speed and resolution in other parts of the human anatomy, smaller diameter gradient coils have been used, e.g., smaller diameter head or wrist coils. Typical head coils are on the order of 30 cm in diameter and wrist coils are smaller yet.

[0005] One major drawback of whole body coils, insertable coils, and local coils of the general type described above, is that they limit access to the examined patient. The gradient coils substantially surround the examined region. In order for a physician to gain access to the examined region, such as to perform a biopsy or other interventional procedure, the patient must be removed from within the gradient coil assembly. Moving the patient relative to the gradient coil assembly also moves the patient relative to the resultant image obtained from the coil assembly. The moved patient needs then to be re-registered with the diagnostic image before performing any interventional procedures.

[0006] An alternative coil design having a plurality of access ports defining holes through the gradient coil has been proposed in U.S. Patent No. 5,304,933 to Vavrek, et al. In the Vavrek, et al. system, the local gradient coil is adapted for use with a stereotaxic device and includes an opening in the coil form positioned to minimize the destruction of the gradient fields and a mechanical slide bearing for moving the form with respect to the stereotaxic frame so that the opening may be limited in area and yet provide essentially unrestricted linear access to the patient. The windings in the neighbourhood of the opening are diverted by modifying the stream function of the windings in a manner to minimize the effect of the opening on the resultant gradient field. However, one problem with the Vavrek, et al. system is that a significant level of undesirable torque is generated when the coil is operated in a uniform static magnetic field. In addition, the gradient coil layout in the Vavrek, et al. system is designed or specified using a "forward approach method," thus realizing a coil configuration having a desirable asymmetric current distribution but capable of generating only marginally acceptable levels of gradient strength and slew rate.

[0007] As shown in the Vavrek, et al. patent drawings, the access holes are quite small and therefore provide only modest access to the patient within the coil. The Vavrek, et al. coil design does not allow for building both the X and Y transverse coils on the same coil carrying member radius for improved linearity.

[0008] Another alternative approach to the prior closed gradient coil systems is suggested in U.S. Patent No. 5,378,989 to Barber, et al. This configuration proposes a pair of axially spaced apart cylindrical gradient coil carrying members separated by a distance defining an interventional access area. A patient is disposed within the bores of the coil carrying members in axial alignment therewith. The Barber. ct al. system is constructed such that large portions of current patterns disposed near the isocentre of the transverse coils are flared radially outward in order to permit an opening at the centre of the magnet and along the axial direction.

[0009] Generally, such an arrangement generates a gradient set which is suitable for interventional applications but has very poor gradient field strength and slew rate performance. In addition, primarily due to the flared current pattern described above, one further disadvantage of the Barber, et al. system is very poor rise time performance. Therefore, the Barber, et al. system cannot be used for real time needle tracking or for any fast imaging techniques because of the poor gradient performance and slow rise time of the overall coil set attributable mainly to the "forward approach method" used to specify the coil conductor layout.

[0010] Morich, et al. in their U.S. Patent No. 5,585,724 assigned to the assignee of the present invention, propose yet another alternative to the prior closed gradient coil systems wherein an axially directed interstitial gap is provided

between a set of coil pairs to define an interventional access area. A coil arrangement of the type described in the Morich, et al. patent is particularly well suited for interventional applications in open magnet systems having horizontally directed main magnetic fields. However, the gradient performance of the Morich, et al. design is somewhat limited in both gradient strength and rise time.

[0011] In accordance with the present invention, a magnetic resonance imaging apparatus is provided. A primary magnet assembly generates a temporally constant magnetic field through an examination region. A gradient coil assembly induces magnetic field gradients across the examination region. A radio frequency coil receives resonance signals from the examination region. The gradient coil carries first and second spaced apart gradient coil portions which are physically separated from each other on a coil carrying member to define at least one azimuthally directed gap extending along the coil carrying member between the first and second gradient coil portions. The first and second gradient coil portions are disposed in the examination region towards opposite top and bottom sides of the examination region. Each of the first and second gradient coil portions include internal windings for generating gradient magnetic field components along at least two mutually orthogonal axes in the examination region.

[0012] According to a preferred feature of the invention, the gradient coil assembly may be a two-part coil assembly such as a system having a hingeable/removable split top coil to facilitate easy patient receipt and exiting from the magnetic resonance imaging system. Further, the top half of the coil can be easily removed or hinged open to provide surgical access. Such access is invaluable when performing interventional MRI using an open magnet with a horizontally directed field.

[0013] Ways of carrying out the invention will now be described in detail, by way of example, with reference to the accompanying drawings, in which:

FIGURE 1 is a diagrammatic illustration of an open magnetic resonance imaging system using an octapole gradient coil system in accordance with the present invention;

FIGURE 2 is an expanded view of the split top octapole gradient coil assembly of FIGURE 1;

FIGURE 3 is an expanded view of the split top coil assembly shown in FIGURE 2 and illustrating a first preferred gradient coil arrangement on the coil carrying members wherein the coils are arranged in a stacked relationship;

FIGURE 4 is a flat illustration of a representative bunched coil pattern of a transverse gradient coil with an azimuthal gap from FIGURE 3;

FIGURE 5 is a schematic illustration of the Z-axis coil according to the present invention constructed on the split top gradient coil assembly of FIGURE 2;

FIGURE 6 is an expanded view of the split top gradient coil assembly shown in FIGURE 2 and illustrating a second preferred gradient coil arrangement on the coil carrying members wherein the coils are constructed to fit upon the same radius;

FIGURE 7 is a flat illustration of a representative bunched coil pattem of transverse gradient coils constructed to fit upon the same radius as in FIGURE 6;

FIGURES 8 and 9 are schematic illustrations of alternative sliding top and hinged top constructions respectively for connecting the split top octapole gradient coil system shown in FIGURE 2;

FIGURE 10 is an illustration of the one-half discrete current distribution generated for an analytical model octapole X-gradient coil in accordance with the present invention;

FIGURE 11 shows experimental measurements of the on-axis behaviour of the octapole X-gradient coil of FIGURE 10 with a 3.5 cm azimuthal gap and a field strength of 24.2 mT/m at 199.53 amps;

FIGURE 12 is a diagrammatic illustration of a "C magnet" type magnetic resonance imaging system using the octapole gradient coil system in accordance with the present invention;

FIGURE 13 is a flat illustration of a representative Y-gradient coil pattern for use with imaging apparatus as in FIGURE 12 having main magnets generating vertically directed fields;

FIGURE 14 is an expanded view illustrating a preferred coil arrangement of the Y-gradient coils of FIGURE 13

disposed on the split top coil carrying member formed in accordance with the present invention;

FIGURE 15 is a flat illustration of a representative X-gradient coil pattern for use with imaging apparatus as in FIGURE 12 having main magnets generating vertically directed fields; and

FIGURE 16 is an expanded view illustrating a preferred coil arrangement of the X-gradient coils of FIGURE 15 disposed on the split top coil carrying member formed in accordance with the present invention.

[0014] With reference to FIGURE 1, an open or "double doughnut" type magnetic resonance imaging system **1** includes main or primary magnets **10**, **11**. Preferably, the main or primary magnets are annual superconducting magnets disposed adjacent opposite ends of the assembly within liquid helium cans **12**, **13**. The liquid helium cans and the magnets are surrounded by a plurality of cold shields **14**, **15** which assist in maintaining the superconducting magnets at superconducting temperatures while minimizing helium boil off. The magnet assemblies are surrounded by a pair of toroidal vacuum dewars **16**, **17**. The magnets **10**, **11** are spaced apart to provide access to the patient. The helium cans, cold shields, and vacuum dewars are preferably separately closed to maintain their integrity.

[0015] The vacuum dewars and the magnet assemblies define a pair of central bores **20**, **21**. A gradient coil assembly, preferably self-shielded whole body gradient coil assemblies **19**, **23** are disposed around an outer periphery of the bore **20**, **21**. In the illustrated embodiment, the whole body gradient coil assemblies **19**, **23** include primary gradient coil assemblies **24**, **25** within the bore which include X, Y, and Z whole body gradient coil windings for generating magnetic field gradients along X, Y, and Z directions. A shield gradient coil assembly **26**, **27** is disposed within the vacuum dewar for cancelling the magnetic field gradients emanating toward the main magnetic assembly. A radio frequency shield **30**, **31** lines an inner surface of the gradient coil assembly. The radio frequency shield is transparent to gradient (kHz) range magnetic fields but is opaque to radio frequency (Mhz) signals. Whole body radio frequency coils **32**, **33**, such as bird cage style coils, are disposed around the inner surface of the radio frequency shield **30**, **31** surrounding the bore **20**, **21**.

[0016] A retractable patient support **39** supports a subject to be examined and an insertable gradient coil assembly **22** formed in accordance with the present invention. The patient support includes a supporting surface in a substantially horizontal plane as shown. The supporting surface has a longitudinal axis lengthwise along the surface and a perpendicular transverse axis across the surface, both in the horizontal plane. The supporting surface is slidably mounted on a patient support frame to provide a means for moving the supported surface in the horizontal plane. Preferably, a motor drive (not shown) is mounted to the frame to drive the patient supporting surface along the frame.

[0017] With continuing reference to FIGURE 1 and with further reference to FIGURE 2, the insertable gradient coil assembly **22** includes a lower portion **44** which is mounted to the patient support **39**. An upper, selectively removable portion **46** is electrically and mechanically interconnected with the lower portion **44** for imaging and is released and removed to facilitate patient access. In the preferred embodiment, the upper and lower gradient coil assemblies **46**, **44** are constructed on a rigid dielectric former. The upper and lower assemblies together carry a pair of X-gradient coil assemblies as illustrated in FIGURES 4 and 7 according to a first preferred embodiment and a pair of Y-gradient coil assemblies also as illustrated in FIGURES 4 and 7 according to a first preferred embodiment. In the first preferred embodiment for horizontal main field imaging systems, the X and Y-gradient coils are arranged on the formers substantially as shown in FIGURES 3 and 6. Z-gradient coils in the form of annular loops are also carried by the dielectric formers preferably in a manner as shown in FIGURE 5. In the first and alternative preferred embodiments, the X and Y-gradient coils are constructed of copper foil laminated to the dielectric former. Electrical connectors, such as metal pins and sockets, are mounted in the upper and lower former portions for providing electrical continuity between coil portions on the upper and lower coil portions when the coil portions are assembled together.

[0018] In operation, a sequence control **71** generates the appropriate gradient and radio frequency pulses of a selected magnetic resonance imaging sequence. More specifically, the characteristics of a selected magnetic resonance imaging sequence are withdrawn from a sequence memory **73** and stored and used to control a radio frequency sequence controller **75** and a gradient pulse controller **77**. A common clock **79** clocks the radio frequency and gradient controllers simultaneously. The selected radio frequency pulse signals are conveyed to a radio frequency transmitter **81** which is selectively connectable to the whole body radio frequency coil **33** and to an insertable radio frequency coil.

[0019] After magnetic resonance is induced, the insertable radio frequency coil or a surface coil (not shown) receives the magnetic resonance signal and conveys it to a digital receiver **83**. The digital receiver **83** demodulates and digitizes the magnetic resonance signal. A reconstruction processor **85** reconstructs the received magnetic resonance signals into a volumetric or slice image representation. A volume image memory **87** stores one or a series of image representations from the reconstruction processor. A video processor **89** converts the selected portions of the image representation in the image memory **87** into appropriate format for display on a monitor **91**. For example, the video processor may convert selected slices of an imaged volume into appropriate format for display. As another option, the video processor can select the corresponding slice in a series or temporally displaced images of the heart to provide a cine

image representation which simulates a motion picture of the selected slice of the heart as the heart is beating. As yet another example, the video processor can assemble a three-dimensional rendering of a selected organ or region.

[0020] With continued particular reference to FIGURE 2, the insertable gradient coil assembly **22** includes a bottom half gradient coil portion **44** and a top half gradient coil portion **46** as described generally above. The top and bottom half portions are substantially separated by a pair of left and right axially extending elongate access apertures **48, 50**. The left and right access apertures are preferably circumferentially spaced apart by about 180° on the cylindrical coil assembly **22** as shown. Each of the top and bottom half portions **46, 44** are additionally respectively substantially bisected by a top axially extending elongate aperture **52** and a bottom axially extending elongate access aperture **54**. The top and bottom access apertures are also preferably circumferentially spaced apart by about 180° on the cylindrical coil assembly **22** as shown. The top access aperture **52** divides the top half gradient coil portion **46** into a top left coil portion **56** and a mirror image top right coil portion **58**. Similarly, the bottom access aperture **54** divides the bottom half gradient coil portion **44** into a bottom left coil portion **60** and a mirror image bottom right coil portion **62**. The apertures are formed by a set of azimuthal gaps defining a set of elongate circumferentially spaced apart ports extending along the entire length of the coil assembly to provide an interventionist access to a patient received within the closed coil assembly. Further, the top half portion **46** is electrically and mechanically interconnected with the bottom half portion **44** for imaging and is released and removed to facilitate patient entry and exit from the coil assembly. Preferably, the bottom and top half gradient coil portions **44, 46** are constructed on a rigid dielectric former.

[0021] Turning now to FIGURES 3 and 4, a first preferred transverse gradient coil arrangement according to the present invention will be described. In the first preferred embodiment, each of the bottom and top half gradient coil portions **44, 46** carry four X-gradient coil members and four Y-gradient coil members, respectively. The gradient coil members are constructed from copper sheets which are either machined or etched into the desired patterns as shown. The sheets are preferably bonded to an FR-4 backing material and are arranged on the bottom and top half gradient coil portions substantially as shown in the FIGURE. More particularly, the top half gradient coil portion **46** carries four discrete X-gradient coil members **80-83** generally as shown. In a similar fashion, the bottom half gradient coil portion **44** carries a corresponding set of X-gradient coil members **84-87** in a mirror image substantially as shown. The preferred bunched coil patterns for the first set of X-gradient coil members **80-83** carried on the top half gradient coil portion **46** are shown in FIGURE 4. The other X-gradient coil members **84-87** disposed on the bottom half gradient coil portion **44** are preferably constructed substantially as shown in FIGURE 4 as well.

[0022] With continued reference to FIGURE 3 in particular, the top half gradient coil portion **46** carries four discrete Y-gradient coil members **90-93** generally as shown. In a similar fashion, the bottom half gradient coil portion **44** carries a corresponding set of Y-gradient coil members **94-97** in a mirror image substantially as shown. As shown in FIGURE 3, the Y-gradient coil members **90-97** are stacked onto the X-gradient coil members **80-87**. However, alternatively, the X-gradient coil members may be stacked on the Y-gradient coil members or, as yet another alternative, the stacking arrangement between the X and Y-gradient coil members may be randomly or evenly distributed over the insertable gradient coil assembly **22**.

[0023] Turning now to FIGURE 5, a schematic illustration of the Z-axis coil assembly constructed on the split top gradient coil assembly of FIGURE 2 according to the present invention will now be described. The Z-axis coil **100** includes a top set of Z-axis coil members **102** and a corresponding mirror image bottom set of Z-axis coil members **104**. Electric current flowing through the top and bottom set of Z-axis coil members **102, 104** cooperate to generate the Z-axis gradient field in a well known manner. Preferably, the Z-axis coil is constructed from copper sheets which are either machined or etched into the desired pattern shown. The machined or etched copper sheets are then bonded to an FR-4 backing material and are then rolled or bent into the desired shape shown in FIGURE 5. As would be appreciated by those skilled in the art, the Z-axis pattern could also be constructed by abrasive water jet cutting or by wire winding or the like. Further, although FIGURE 5 illustrates a single reversal in the Z-axis windings occurring at the isocentre of the coil, other designs could be easily accommodated. The returns are routed along left and right flanges **106, 108** as shown so that the field from the top and bottom sets **102, 104** cancel. Although each half of the Z-axis coil will separately generate a net torque, this torque is constrained in the preferred embodiment by alignment pins and latching mechanisms located in the flanges of the gradient coil formers described below in greater detail with reference to FIGURES 8 and 9.

[0024] Turning now to FIGURES 6 and 7, a second preferred transverse gradient coil arrangement for use with imaging equipment generating horizontally directed fields will now be described. In FIGURE 6, each of the bottom and top half gradient coil portions **44, 46** carry four X-gradient coil members and four Y-gradient coil members, respectively. In this second preferred embodiment, both the X and Y-gradient coil members are radially co-located on the gradient coil former in a manner substantially as shown. That is, the coils are constructed to fit upon the same radius. One advantage is an increase in the available patient aperture space. In addition, the radial co-location of the X and Y transverse gradient coil members allows for a decrease in size of the overall gradient coil assembly structure since only a single set of gradient coils need to be constructed rather than two overlapping sets as was traditionally required.

[0025] In the second preferred embodiment shown, the top half coil portion **46** carries four discrete X-gradient coil

members **80'-83'** and four discrete Y-gradient coil members **90'-93'** generally as shown. In a similar fashion, the bottom half gradient coil portion **44** carries a corresponding set of X-gradient coil members **84'-87'** and a set of Y-gradient coil members **94'-97'** in a mirror image substantially as shown. The coil members are preferably constructed from machined or etched copper sheets and then bonded together substantially as described above.

[0026] Overall, the X and Y-gradient coil members are symmetrically spaced apart and disposed over the gradient coil assembly **22** for the purpose of minimizing the effects of net torque on the coil structures. As shown, the coil members experience no net torque effects. This significantly reduces the coil vibration which is normally present in coil structures having non-symmetric current patterns. As a result, the medical image acquired in the magnetic resonance imaging system **10** has only minimum distortion and minimum ghosting effects.

[0027] In addition to the above, since the current distribution of the coil arrangements shown in FIGURES 4 and 7 advantageously permit the coils to be disposed over the entire length of the gradient coil assembly **22**, there are none of the space limitations that are typically encountered in prior art gradient coil systems of the type described above by way of background. As a direct result, the required stored magnetic energy of the present invention is substantially lower than the required stored magnetic energy in prior art systems. As TABLE 3 below indicates. the octapole gradient coil with azimuthal gap of arbitrary width in accordance with the present invention has double the slew rate on 30% reduced length as compared against the prior art coil assembly of Morich, et al. In addition, the gradient coil arrangements according to the present invention generate six times the gradient strength and two hundred times the slew rate of the Vavrek, et al. open gradient coil system described generally above.

[0028] Turning now to FIGURES 8 and 9, alternative preferred embodiments for connecting the split top octapole gradient coil system according to the present invention are shown. First, with regard to FIGURE 8, the bottom half gradient coil portion **44** is adapted for connection to the patient supporting surface **36** as described above and further includes a pair of left and right connection flanges **110, 112**. The left and right connection flanges are adapted to slidingly engage and interlock with a corresponding set of left and right connection flanges **114, 116** formed on the top half gradient coil portion **46**. The left and right connection flanges **114, 116** are adapted to lock onto the bottom left and right connection flanges using any suitable mechanical connecting means such as the hook members **118, 120** as shown. The hook members extend vertically past the left and right connection flanges **110, 112** and engage the back sides thereof in a manner substantially as shown. Of course, alternative sliding connection arrangements are possible as would be understood by those skilled in the art.

[0029] Turning to FIGURE 9, an alternative split top octapole gradient coil connecting system is shown including a pair of left and right connection hinges **122, 124** which are adapted to connect the bottom and top half gradient coil portions **44, 46** in a manner substantially as shown. As described above, the bottom half gradient coil portion **44** is connected preferably to the patient supporting surface **36**. The left and right hinges enable the top half gradient coil portion **46** to open in a manner as shown to permit easy patient entrance and removal from the gradient coil assembly **22**.

[0030] In each of the alternative preferred split top octapole gradient coil system connecting arrangements shown in FIGURES 8 and 9, a plurality of electrical contacts **130-131** are provided on the connection flanges **110-113** in a manner to electrically connect the gradient coil members disposed on the top half gradient coil portion **46** to those disposed on the bottom half gradient coil portion **44**. In the embodiment illustrated in FIGURE 9, the electrical connection elements are electric contact strips for slidable mechanical and electrical engagement between the corresponding opposed contact members. In the embodiment shown in FIGURE 9, the electrical contact elements are preferably pin and socket members which are adapted for intermateable connection when the top half gradient coil portion **46** is hinged into a closed position with the bottom half gradient coil portion **44**.

[0031] The novel preferred method of generating a discrete transverse current density with coils arranged to define an azimuthal gap of arbitrary width will next be described. The technique is based in part on the so-called Turner energy minimization approach.

[0032] According to the present invention, an analytical expression for the azimuthal component of the current density, J_ϕ^a, is found that both satisfies all the predetermined symmetry conditions which are in accord with the symmetry behaviour of traditional transverse gradient coil geometry, and vanishes on any predetermined azimuthal (ϕ') location. With regard to the symmetry conditions, the current density J_ϕ^a is symmetric along the z direction. The current density J_ϕ^a is also symmetric along the z direction around the isocentre of the coil ($\phi = 0°$). Therefore, the appropriate expression for J_ϕ^a is:

$$J_\phi^a = \cos\phi\,[\cos\phi - \cos\phi'] \sum_{n=1}^{\infty} j_n^a \cos(k_n z) \quad for\ |z| \le \frac{L}{2} \qquad (1)$$

where j_n^a are the Fourier expansion coefficients in the expression, L is the total length of the coil, $k_n = (2n\pi/L)$, and ϕ' represents the angle whereby J_ϕ^a vanishes. Using the continuity equation ($\vec{\nabla} \cdot \vec{J} = 0$), the expression for the axial com-

ponent of the current density is:

$$J_z^a = \sin\phi\,[2\cos\phi - \cos\phi'] \sum_{n=1}^{\infty} \frac{j_n^a}{k_n a} \quad for\ |z| \leq \frac{L}{2} \tag{2}$$

where *a* is the radius of the coil in the expression. Both currents are zero for |z| ≥ *L*/2.

[0033] The expression of the axial component (z) of the magnetic field for any angular dependence and at a radial position which is less than the radius *a* of the coil is:

$$B_z = -\frac{\mu_o a}{2\pi} \sum_{m=-\infty}^{m=\infty} \int_{-\infty}^{\infty} dk\, e^{im\phi + ikz} k J_\phi^a(m,k)\, I_m(k\rho)\, K_m(ka)\, S_{ma} \tag{3}$$

where $I_m(k\rho)K'_m(ka)$ are the modified Bessel functions of the first and second order, and

$$S_{mab} = 1 - \frac{I'm(ka)\ K'm(kb)}{I'm(kb)\ K'm(ka)} \tag{4}$$

when shielding coil with radius *b* is present. S_{mab} when only a primary coil is present.

[0034] The expression for the magnetic energy stored in the coil is:

$$W_m = \frac{\mu_o a^2}{2} \sum_{m=-\infty}^{m=+\infty} \int_{-\infty}^{\infty} dk\, I_m'(ka)\, K_m'(ka)\, S_{mab} |J_\phi^a(m,k)|^2 \tag{5}$$

[0035] Considering the Fourier transform of the current density $J_\phi^a\,(m, k)$

$$j_\phi^a(m,k) = \frac{L}{2} \sum_{n=1}^{\infty} \psi_n(k)\, j_n^a \cdot \begin{cases} \frac{1}{2}\ for\ m=0 \\ -\frac{1}{2}\cos\phi'\ for\ m=1 \\ \frac{1}{4}\ for\ m=2 \end{cases} \tag{6}$$

where

$$\psi_n(k) = \left[\frac{\sin(k-k_n)\frac{L}{2}}{(k-k_n)\frac{L}{2}} + \frac{\sin(k+k_n)\frac{L}{2}}{(k+k_n)\frac{L}{2}} \right] \tag{7}$$

[0036] Therefore the expression of the magnetic field becomes:

$$B_z = -\frac{\mu_0 aL}{4\pi}\sum_{n=1}^{\infty} j_n^a \int_0^{\infty} dk\, k\psi_n(k)\cos kz\,[I_0(k\rho)K_0'(ka)S_{0ab} + \cos 2\phi\, I_2(k\rho)K_2'(ka)S_{2ab} - 2\cos\phi\cos\phi' I_1(k\rho)K_1(ka)S_{1ab}] \tag{8}$$

[0037] The expression of the magnetic energy stored in the coil is:

$$W_m = -\frac{\mu_0 a^2 L^2}{16}\sum_{n=1}^{\infty}\sum_{n'=1}^{\infty} j_n^a j_{n'}^a \int_0^{\infty} dk\,\psi_n(k)\psi_n'(k)\,[I_0'(ka)K_0'(ka)S_{0ab} + 2\cos^2\phi' I_1'(ka)K_1'(ka)S_{1ab} + \frac{1}{2}I_2'(ka)K_2'(ka)S_{2ab}] \tag{9}$$

[0038] Using a standard energy minimization procedure, preferably the Turner approach, the functional ε in terms of the stored magnetic energy and the magnetic field is constructed as:

$$\epsilon(J_n^a) = W_m - \sum_{j=1}^{N}\lambda_j\left(B_z(\vec{r}_j) - B_{zSC}(\vec{r}_j)\right) \tag{10}$$

where N represents the total number of constraint points in the expression and $B_{zSC}(\vec{r}_j)$ is the value of the gradient field at the constrain point $\vec{r}_j$. Minimizing ε with respect to $j_{n''}^a$, a matrix equation for $j_{n''}^a$ is obtained as:

$$\sum_{n'=1}^{\infty} j_{n'}^a \left\{\frac{aL\pi}{2}\int_{-\infty}^{\infty} dk\,\psi_n(k)\psi_{n'}(k)\right\} = \sum_{j=1}^{N}\lambda_j \int_{-\infty}^{\infty} dk\, k\cos kz_j\psi_n(k) \tag{11}$$

[0039] Equivalently, the matrix equation for the current density j_n^a is:

$$\sum_{n'=1}^{\infty} j_{n'}^a C_{n'n} = \sum_{j=1}^{N}\lambda_j D_{jn} \tag{12}$$

[0040] Truncating the infinite summations at M terms. the matrix representation of the previous equation (12) becomes:

$$\underline{J}^a\underline{C} = \underline{\lambda}\,\underline{D} \text{ or } \underline{J}^a = \underline{\lambda}\,\underline{D}\,\underline{C}^{-1} \tag{13}$$

where $\underline{J}^a$ is a 1 x M matrix, $\underline{C}$ is a M x M matrix, $\underline{\lambda}$ is a 1 x N matrix and $\underline{D}$ is a N x M matrix in the expression.

[0041] The Lagrange multipliers are found using the expression of the magnetic field. The matrix representation of

the magnetic field thus becomes:

$$B_z(r_j) = \sum_{n=1}^{M} j_n^a D_{jn} \text{ or } \underline{B}_z = \underline{J}^a \underline{D}^t \qquad (14)$$

where $\underline{B}_z$ is a 1 x N matrix in the expression and the superscript t is the symbol for the transpose matrix. Replacing $\underline{J}^a$ from the equation (13) to equation (14) above, the expression for the magnetic field becomes:

$$\underline{B}_z = \lambda \underline{DC}^{-1}\, \underline{D} \qquad (15)$$

which leads to the determination of Lagrange multipliers as:

$$\underline{\lambda} = \underline{B}_z [\underline{DC}^{-1}\, \underline{D}^t]^{-1} \qquad (16)$$

providing that the inverse matrix for the expression $[\underline{DC}^{-1}\underline{D}^t]$ exists. Upon determination of Lagrange multipliers, the matrix form expression for the Fourier components of the current density is:

$$\underline{J}^a = \underline{B}_z [\underline{DC}^{-1}\, \underline{D}^t]^{-1}\, \underline{DC}^{-1} \qquad (17)$$

[0042] In the present invention, the expression for the Fourier components of the current for the gradient coil are solved in the straightforward manner described above. The continuous distribution of the current density for the coil is generated by substituting the Fourier components of the current back into the expression of the current density J_ϕ^a.

[0043] With reference next to FIGURES 10 and 11, the design of an analytical model construction of an Octapole X-gradient coil with the azimuthal gap according to the present invention will be described. The model assumes a cylindrical radius of 191.3 mm and a total length of 60 cm. The total length of the azimuthal gap was arbitrarily selected to be 3.5 cm (ϕ' = 10°). Three constraint points were next selected to specify the quality of the gradient field inside a 23 cm DSV.

[0044] With reference to TABLE 1 below, the first constraint point sets the strength of gradient field to 24.5 mT/m, the second limits the variation of the gradient field along its gradient axis (X) to be within 15% from its ideal value at distance of 0.115m from the geometric centre of the coil, and the third constraint defines a 14% gradient field uniformity inside the 23 cm DSV.

TABLE 1

n	P_i	Z_i	$B_{zsc}(n)$
1	0.0010	0.000	0.000024530
2	0.1150	0.0000	0.003127600
3	0.0010	0.1150	0.00002085050

[0045] Solving the inverse problem using the specific set of constraint points set forth above, the continuous current distribution for the X-gradient coil are generated. Applying the stream function technique, the discrete current pattern is obtained as shown in FIGURE 11. The common value of the current for the 12 discrete loops is 199.53 Amps.

[0046] As a next step, the discrete coil pattern is used to calculate the magnetic field via the Biot-Savart law. Thus, it is ensured that the discretization mechanism is appropriate and the choice of the number of discrete loops is adequate to generate the desired quality of the magnetic field.

[0047] The gradient field of the discrete coil pattern is very linear along its primary gradient axis (X). The net gradient field strength at the origin is 24.53 mT/m which is a 0.1% deviation from its constrained value. The linearity of the coil inside the 23 cm DSV is 17.89%.

[0048] FIGURE 11 shows the level of uniformity of the gradient field inside the 23 cm DSV. Specifically, for the 12 discrete loop configuration example described here, the overall uniformity of the gradient inside the 23 cm DSV is 14%.

TABLE 2 below illustrates all the properties of the octapole X-gradient coil as well as a comparison against a X-gradient coil with interstitial gap that has the same gradient strength. Comparing both gradients, it is clear that for an input current/voltage of 300A/400V, the slew rate of the octapole gradient coil with the azimuthal gap according to the present invention is twice as fast as the slew rate of a typical prior art X-gradient coil with the interstitial gap.

TABLE 2

Gradient Property at 300A/400V		
Property	OCTAPOLE COIL	INTERSTITIAL GAP COIL
Radius	191.3 mm	191.3 mm
Length	60 cm	90 cm
Gap	3.5 cm	15 cm
GRADIENT Str. (300 A)	36.89 mT/m	36.89 mT/m
Energy (300 A)	7.28 J	15.35 J
Linearity (23 cm DSV)	17.49%	8.55%
Uniformity (23 cm DSV)	14%	27%
Inductance	162 μH	341 μH
No. Loops	12	16
Resistance	0.041 Ω	0.055 Ω
Linear Rise Time (440 V)	125 μsec	267 μsec
SLEW RATE	294 mT/m/msec	138 mT/m/msec

[0049] Based on the discrete pattern of FIGURE 10, a physical prototype X-gradient coil was constructed by applying a 12 AWG insulated wire onto a fibreglass cylindrical tube. The coil interconnects were placed on the tube in a manner that they would not interfere with the azimuthal gap. The measured inductance of the prototype coil was 158μH at 1 kHz, which represents only a 2.5% deviation from the theoretical value. Using a search coil, the gradient field is plotted along its primary X-gradient axis as shown in FIGURE 14. Furthermore, TABLE 3 below shows a comparison of the gradient field (mT/m/A) between the numerical and the experimental results at various points along the major X-gradient coil axis. It is clear that there is a very good agreement between the analytical and the measured values of the gradient field.

TABLE 3

Position	Analytical	Experimental	Percentage Difference
2.5 cm	0.1227 mT/m/A	0.1204 mT/m/A	+1.8% *
5.0 cm	0.1202 mT/m/A	0.1223 mT/m/A	-1.7% *
7.5 cm	0.1162 mT/m/A	0.1184 mT/m/A	-1.9% *
10.0 cm	0.1092 mT/m/A	0.1119 mT/m/A	-2.5% *
12.5 cm	0.1023 mT/m/A	0.10338 mT/m/A	-1.0% *

* Indicates that the percentages are taken with respect to the theoretical values.

[0050] With reference next to FIGURE 12, an alternative preferred embodiment of the octapole gradient coil assembly of the present invention is shown adapted for use in an open "C" magnet type system. The "C" magnet type magnetic imaging system **10'** includes a magnetomotive force means **A** in the form of a pair of resistive magnet drivers **12'**, **14'** disposed adjacent pole pieces **16'**, **18'** on opposite ends of a C-shaped ferromagnetic flux path **B**. The magnetomotive force means **A** together with a magnetic flux stabilizing means **C** creates and stabilizes a magnetic field across an air gap **20'** between the pole faces and along the ferromagnetic flux path **B**. A retractable patient support **D** selectively supports a patient or subject to be examined, together with an insertable gradient coil **22'** in accordance with the present invention in the air gap **20'** defined between the pole faces. An energizing and calibration system **E** is used to set up the magnetic field across the air gap **20'**. Magnetic resonance electronics **F** selectively induce magnetic resonance of dipoles in the image region and process resultant received magnetic resonance signals to create an image or other

diagnostic information.

[0051] The ferromagnetic flux path **B** includes a C-shaped ferromagnetic member **24'** having a first end **26'**, a second end **28'**, and a middle portion **30'**. The first pole piece **16'** on the first end **26'** of the flux path together with the second pole piece **18'** on the second end **28'** define the air gap **20'** therebetween. The C-shaped member **24'** is configured to minimize the length of the ferromagnetic flux path while spacing the ferromagnetic flux path sufficiently from the gap to minimize distortion to the magnetic field in the air gap.

[0052] The magnetic flux stabilizing means **C** includes a superconducting cryodriver **32'** which encircles a segment of the middle portion **30'** of the C-shaped ferromagnetic member **24'**. As is well known by those skilled in the art, the cryodriver **32'** includes a cryostat **34'** which houses an annular superconductor magnet that encircles the flux path.

[0053] The patient support **D** includes a patient supporting surface **36'** in a substantially horizontal plane. The supporting surface has a longitudinal axis lengthwise along the surface and a perpendicular transverse axis across the surface, both in the horizontal plane. An elevation adjusting means **38'** selectively adjusts the relative height of the supporting surface. The supporting surface is slidably mounted on a support frame **40'** to provide a means for moving the supporting surface in the horizontal plane. Preferably, a motor drive **42'** is mounted to the frame to drive the patient supporting surface along the frame.

[0054] Lastly, in connection with FIGURE 12. the electronics section **F** includes a radio frequency transmitter means **60'** which selectively applies radio frequency pulses to a radio frequency coil (not shown) to excite magnetic resonance of dipoles in the gap magnetic field. A receiver **62'** receives magnetic resonance signals from the region of interest using a radio frequency coil (not shown) as an antenna. A gradient coil control **64'** applies electrical pulses to a gradient field coil (not shown) to cause gradients across the gap magnetic field to encode the magnetic resonance signals. An image reconstruction processor **66'** performs an inverse two-dimensional Fourier transfer or other known algorithm to reconstruct an image representation from the received magnetic resonance signals. The image representations are stored in a memory **68'**, displayed on a video monitor **70'**, further processed, communicated to another apparatus, or the like. A central magnetic resonance controller **72'** controls the excitation power control **74'** to the resistive drivers **12'**, **14'**, the RF transmitter **60'**, and the gradient field control **64'** to implement a preselected magnetic resonance imaging sequence as is conventional in the art.

[0055] With continued reference to FIGURE 12 and with additional reference to FIGURES 2, 14, and 16, the insertable gradient coil **22'** is preferably formed of a bottom half gradient coil portion **44'** fixedly attached to the supporting surface and a top half gradient coil portion **46'** removable from the bottom half to facilitate patient receipt and exiting from the magnetic resonance imaging system **10'**. A pair of left and right elongate apertures **48'**, **50'** extend axially along the length of the insertable gradient coil **22'** generally as shown to provide access to a patient received within the closed gradient coil assembly during interventional procedures or the like. In addition. a pair of top and bottom elongate apertures **52'**, **54'** are formed in the insertable gradient coil **22'** generally as shown in accordance with the present invention to further facilitate access to the patient or specimen during interventional procedures in the magnetic resonance imaging system **10'**. For ease of construction, the left and right apertures **48'**, **50'** are formed on the separation or parting line between the top and bottom half gradient coil portions **46'**, **44'**. However, in accordance with the present invention, the apertures can be disposed in any circumferentially spaced apart manner on the gradient coil **22'**.

[0056] The top and bottom half gradient coil portions **46'**, **44'** together carry a set X-gradient coil assemblies as illustrated in FIGURES 13 and 14 and a set of Y-gradient coil assemblies as illustrated in FIGURES 15 and 16. In the second preferred embodiment for use with imaging systems having vertically oriented main fields, the X and Y-gradient coils are arranged on the formers of the insertable gradient coil assembly substantially as shown in FIGURES 14 and 16. The Z-gradient coil is in the form of annular loops carried by the dielectric formers substantially in the manner as shown above in connection with FIGURE 5. In the alternative preferred embodiment illustrated in FIGURES 13-16, the X and Y-gradient coils are constructed of copper foil laminated to the dielectric former. Electrical connectors such as metal pins and sockets, are mounted in the upper and lower former portions as described above, for providing electrical continuity between coil portions on the upper and lower coil portions when the coil members are assembled together.

[0057] With continued reference to FIGURES 13-16, each of the top and bottom half gradient coil portions **46'**, **44'** carry four X-gradient coil members and two Y-gradient coil members, respectively. The Y-gradient coils are configured substantially as shown in FIGURE 13 and are arranged on the former member as shown in FIGURE 14. FIGURE 13 illustrates a pair of Y-gradient coils. Two pairs of Y-gradient coils are carried on each of the top and bottom coil portions **46'**, **44'**.

[0058] The X-gradient coils in the alternative preferred embodiment for use in vertically directed main magnetic fields is shown in FIGURE 15. The X-gradient coils are disposed on the former member substantially as shown in FIGURE 16. In that regard, each of the top and bottom half gradient coil portions **46'**, **44'** carry a pair of X-gradient coil members as shown.

[0059] One advantage of the octapole magnetic resonance gradient coil system with elongate azimuthal gap is an improvement in gradient field strength and slew rate whereby the resolution of the imaged subject is greatly improved. Another advantage resides in improved imaging speeds and reduced data acquisition times. Another advantage is that

it facilitates access to the examined region of the patient,.while the patient is disposed in a known relationship to the gradient magnetic field coils. Yet another advantage is the construction of a gradient magnetic field coil assembly in a snap together shell arrangement whereby the top shell is removable from the bottom shell to facilitate patient receipt and exiting from the magnetic resonance imaging apparatus. Still yet another advantage is an improvement in the linearity of the gradient magnetic field coil assembly due to the transverse coils being disposed at equal radius distances on the coil carrying assembly.

Claims

1. Magnetic resonance imaging apparatus including a primary magnet assembly (10, 11) for generating a temporarily constant magnetic field through an examination region (20), a gradient coil assembly (19, 22, 23) for inducing magnetic field gradients across the examination region (20), and a radio frequency coil (32, 33) for receiving resonance signals from the examination region, the gradient coil assembly including: a coil carrying member (44, 46) carrying first and second spaced apart gradient coil portions (56, 58, 60, 62) which are physically separated from each other on the coil carrying member to define at least one azimuthal gap (48, 50, 52, 54) extending through the coil carrying member and between said first and second gradient coil portions, the first and second coil portions being disposed in the examination region (20) and having internal windings (80-87, 90-97) therein for generating gradient magnetic field components along at least two mutually orthogonal axes (x, y, z) in the examination region (20).

2. Magnetic resonance imaging apparatus as claimed in claim 1, wherein: the coil carrying member is substantially circularly cylindrical defining a first longitudinal axis (Z) and having an outer first circular radius; the first gradient coil portion (56, 58) is substantially circularly semi-cylindrical; the second gradient coil portion (60, 62) is substantially circularly semi-cylindrical; and the at least one azimuthal gap (48, 50, 52, 54) extends along said first longitudinal axis (Z) of the coil carrying member.

3. Magnetic resonance imaging apparatus as claimed in claim 1 or claim 2, wherein: the first gradient coil portion (56, 58) defines a top circularly semi-cylindrical shell portion (46) of said gradient coil assembly and having internal windings (80-83, 90-93) therein for generating a first portion of said gradient magnetic field components along said at least two mutually orthogonal axes (x, y) in the examination region (20); and the second gradient coil portion (60, 62) defines a bottom circularly semi-cylindrical shell portion (48) of said gradient coil assembly being intermateable with the top shell portion and having internal windings (84-87, 94-97) therein for generating a second portion of said gradient magnetic field components along said at least two mutually orthogonal axes (x, y) in the examination region, the top circularly semi-cylindrical shell portion being selectively removable to facilitate patient receipt and exiting from the magnetic resonance imaging apparatus..

4. Magnetic resonance imaging apparatus as claimed in any one of claims 1 to 3, wherein: said internal windings in the first gradient coil portion (56, 58) include a first set of X-axis coils (80-83) for generating a first X-axis component of said gradient magnetic field components and a first set of Y-axis coils (90-93) for generating a first Y-axis component of said gradient magnetic field components; and said internal windings in the second gradient coil portion (60, 62) include a second set of X-axis coils (84-87) for generating a second X-axis component of said gradient magnetic field components and a second set of Y-axis coils (94-97) for generating a second Y-axis component of said gradient magnetic field components.

5. Magnetic resonance imaging apparatus as claimed in claim 4. wherein: said first set of X-axis coils (80-83) of the first gradient coil portion (56, 58) are disposed directly onto said outer first circular radius of said coil carrying member (44, 46) without overlapping said first set of Y-axis coils (90-93) of the first gradient coil portion; and said second set of X-axis coils (84-87) of the second gradient coil portion (60, 62) are disposed directly onto said outer first circular radius of said coil carrying member (44, 46) without overlapping said second set of Y-axis coils (94-97) of the second gradient coil portion.

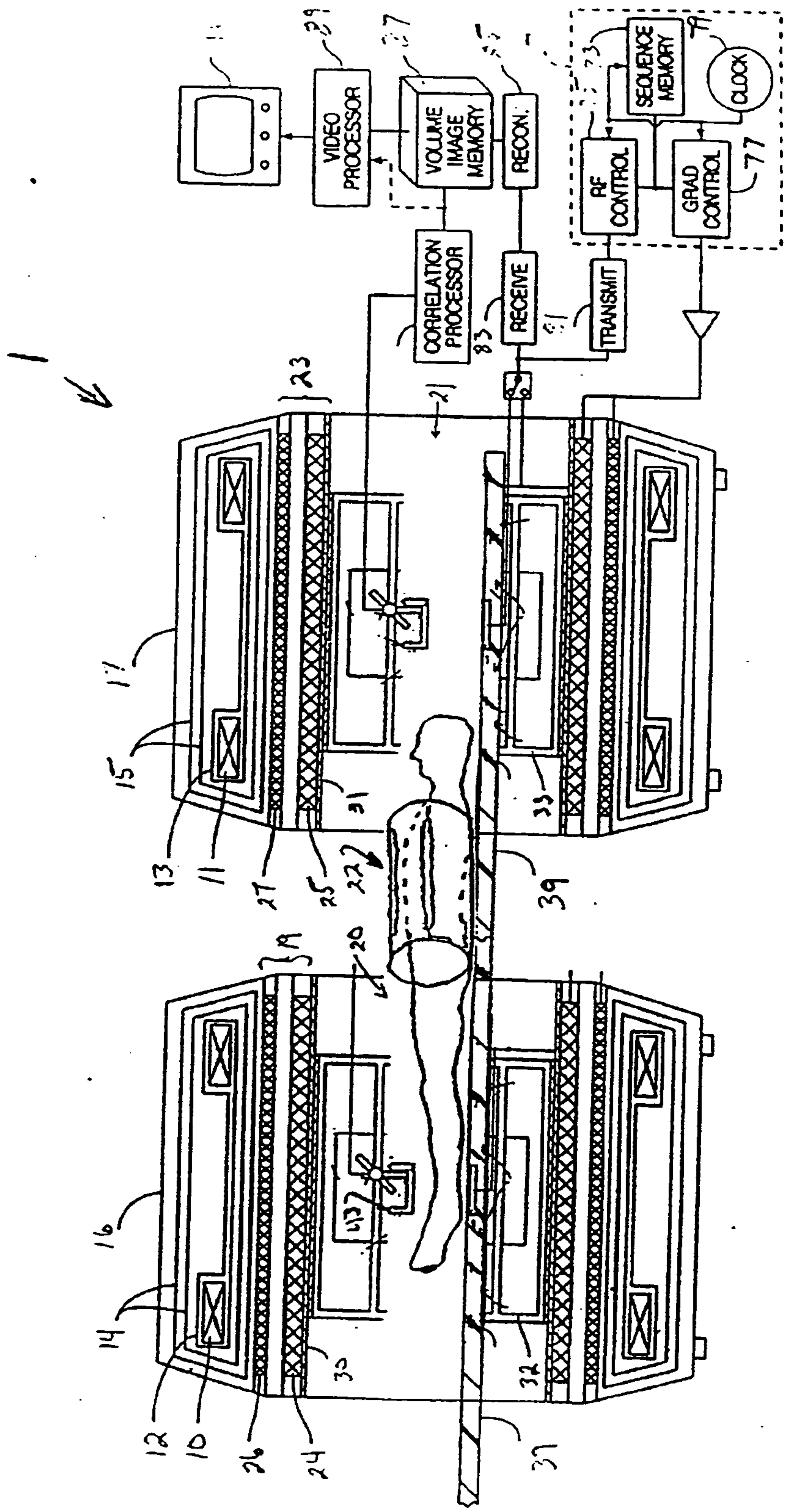
6. Magnetic resonance imaging apparatus as claimed in claim 4, wherein: one of said first set of X and Y-axis coils (80-83, 90-93) of the first gradient coil portion are disposed on said coil carrying member overlapping said the other of said first set of X and Y-axis coils (80-83, 90-93) of the first gradient coil portion; and one of said second set of X and Y-axis coils (84-87, 94-97) of the second gradient coil portion are disposed on said coil carrying member overlapping the other of said second set of X and Y-axis coils (84-87, 94-97) of the second gradient coil portion.

7. Magnetic resonance imaging apparatus as claimed in any one of claims 1 to 6, wherein: said first and second gradient coil portions are physically spaced apart from each other on the coil carrying member to define a plurality of longitudinally extending azimuthally directed gaps (48, 50, 52, 54) spaced apart around said substantially circular cylindrical coil carrying member.

8. A method of magnetic resonance imaging using the apparatus of claim 1.

9. A method of magnetic resonance imaging using the apparatus of any of claims 1 to 7 further including the step of: selectively separating said first and second spaced apart gradient coil portions to facilitate patient receipt and exiting from the magnetic resonance imaging apparatus.

10. A method of magnetic resonance imaging according to claims 8 or 9, further including the step of: performing said magnetic resonance imaging in conjunction with an associated interventional procedure to a patient on the apparatus.


VIDEO PROCESSOR
VOLUME IMAGE MEMORY
RECON.
SEQUENCE MEMORY
CLOCK
RF CONTROL
GRAD CONTROL
CORRELATION PROCESSOR
RECEIVE
TRANSMIT
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39
37
13
11
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32

FIG. 1

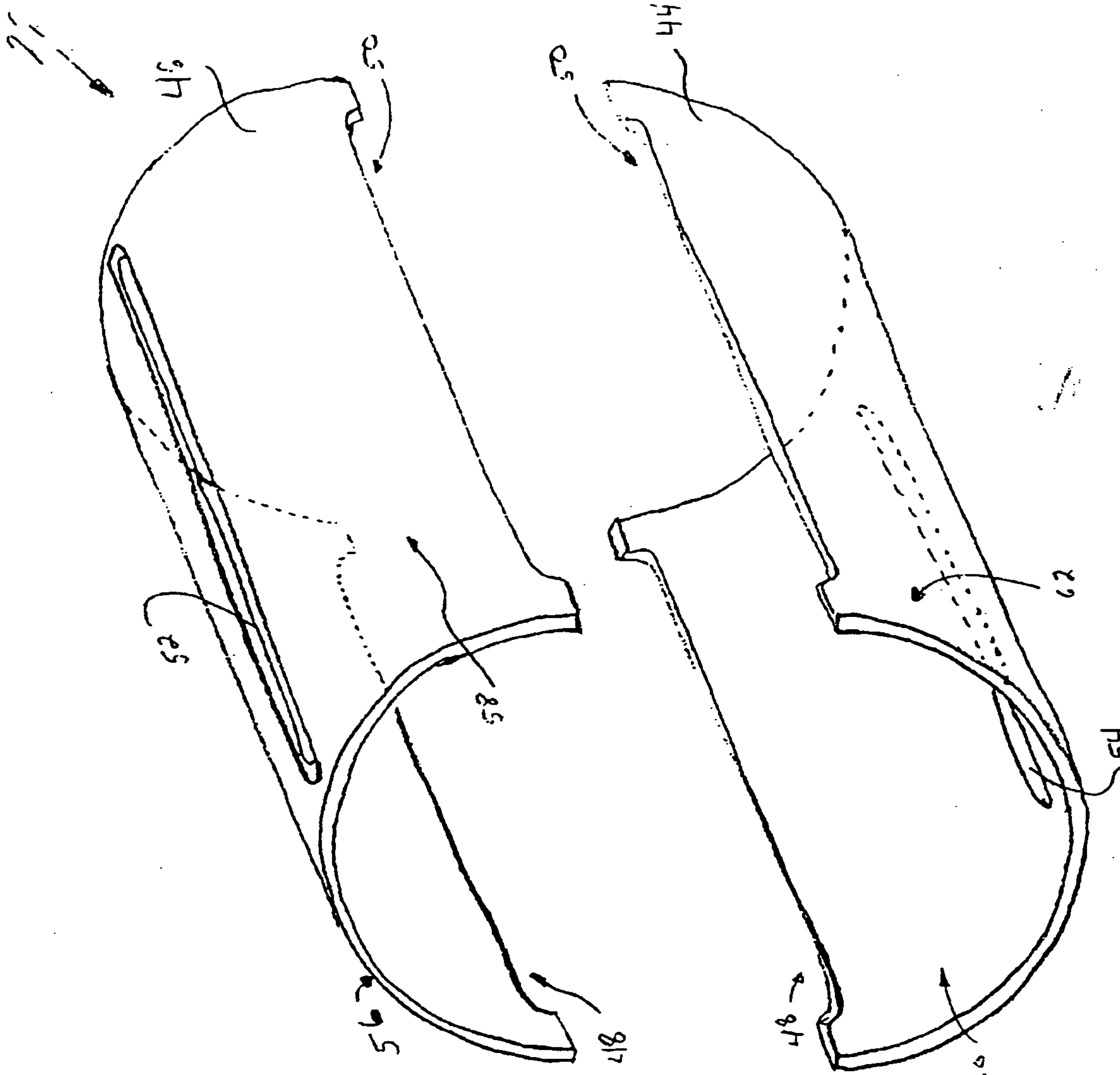
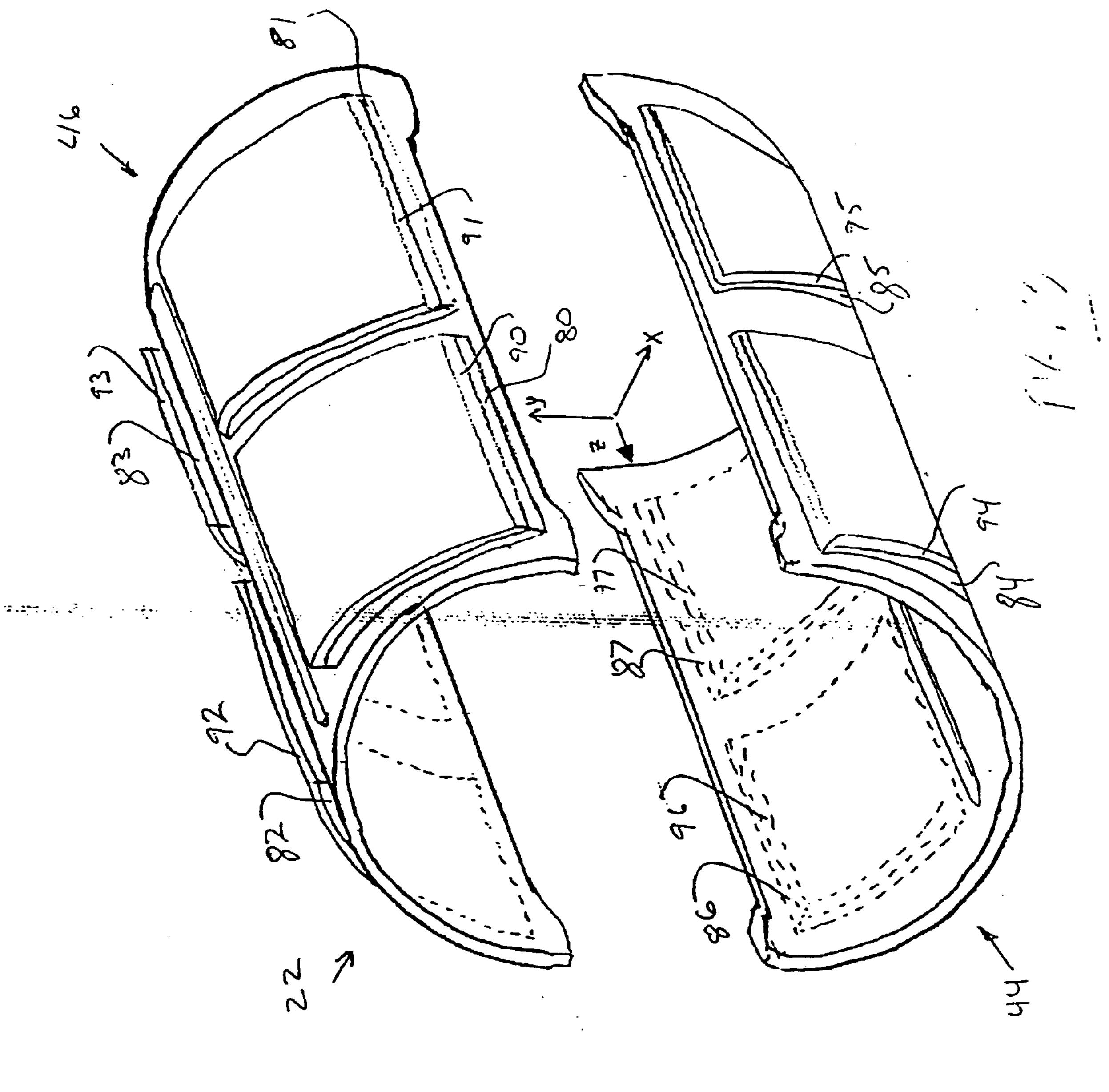


FIG. 2


46
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91
90
80
73
83
72
82
22
75
85
94
84
77
87
96
86
44
x
y
z

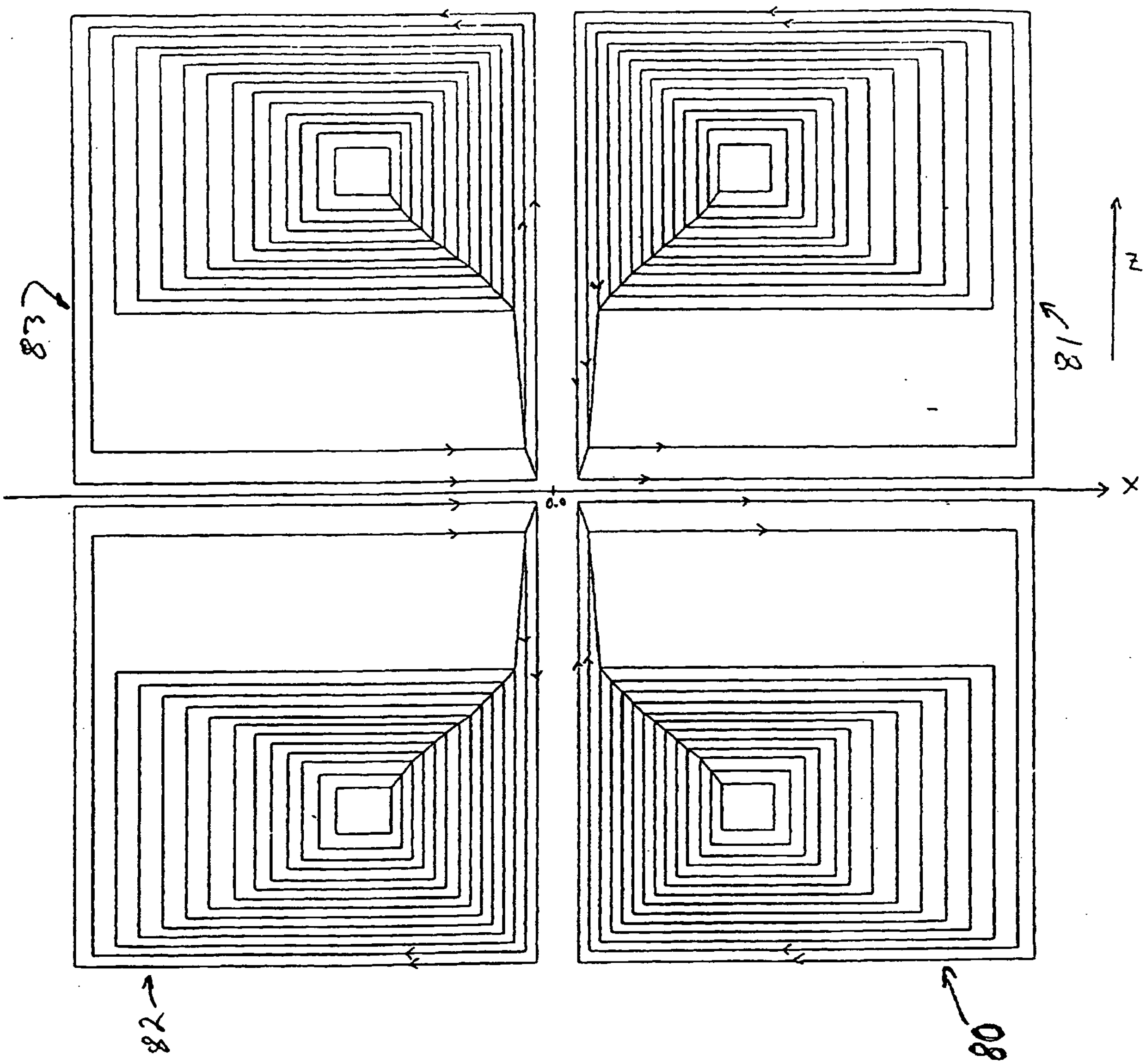


FIGURE 4

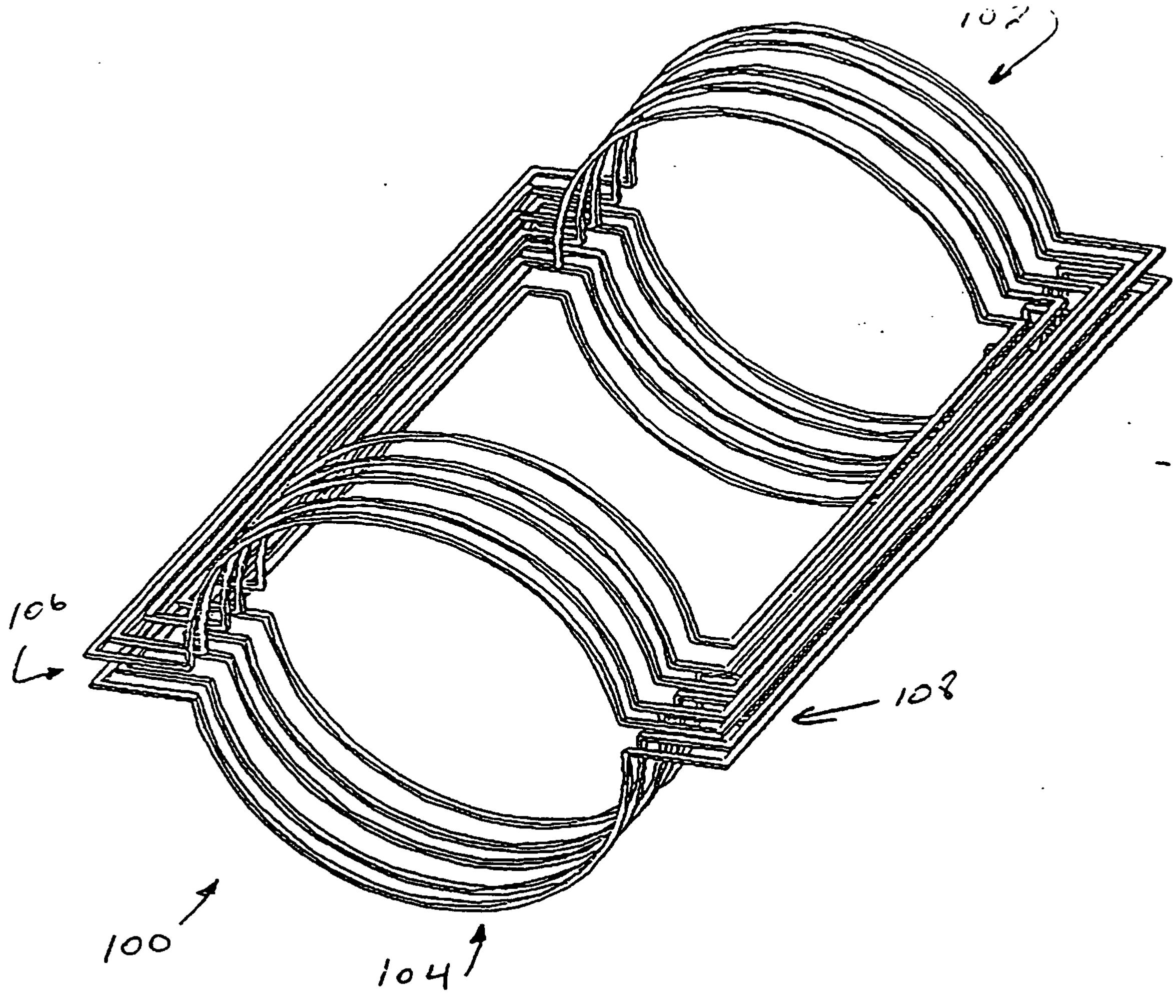


FIG. 5 Z AXIS

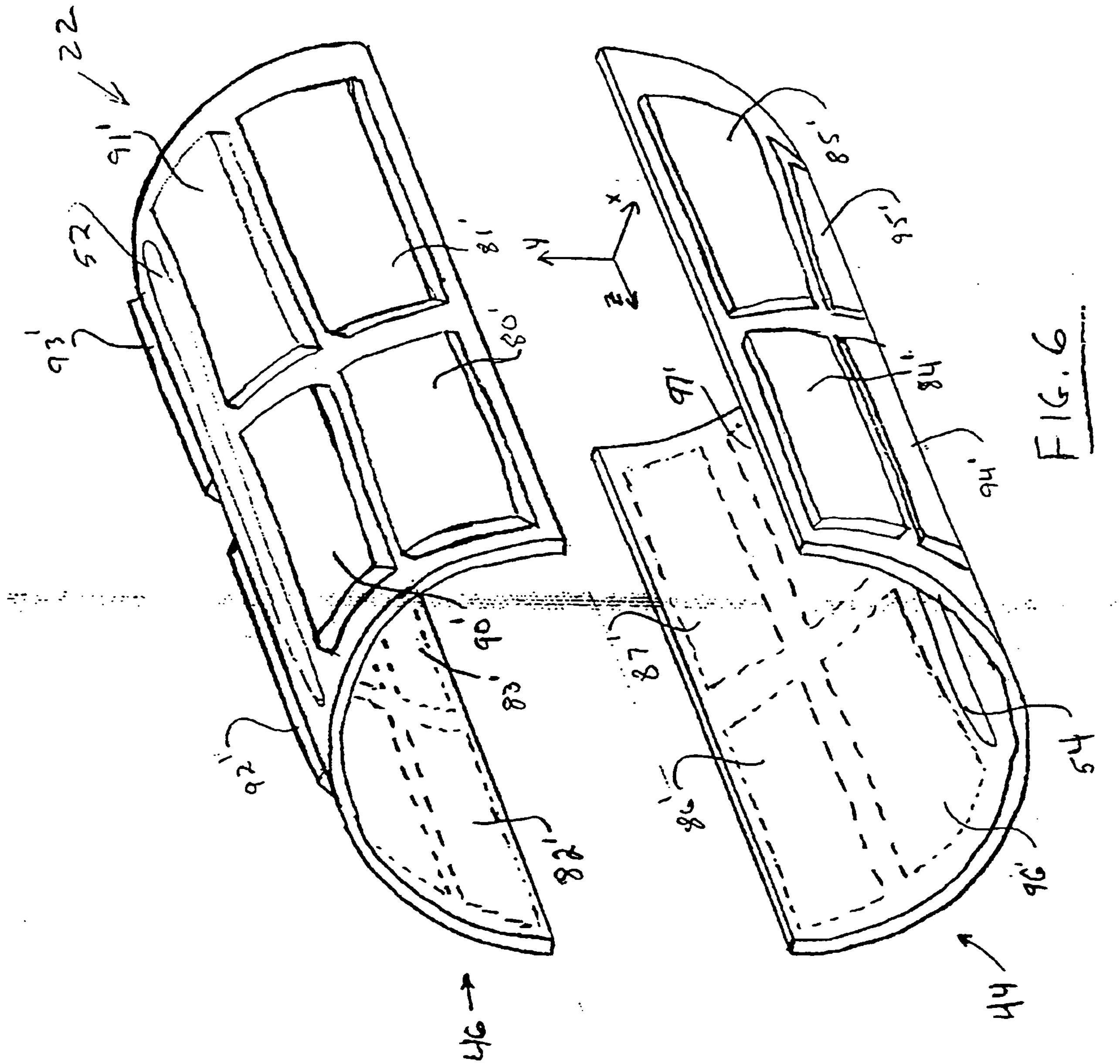


FIG. 6

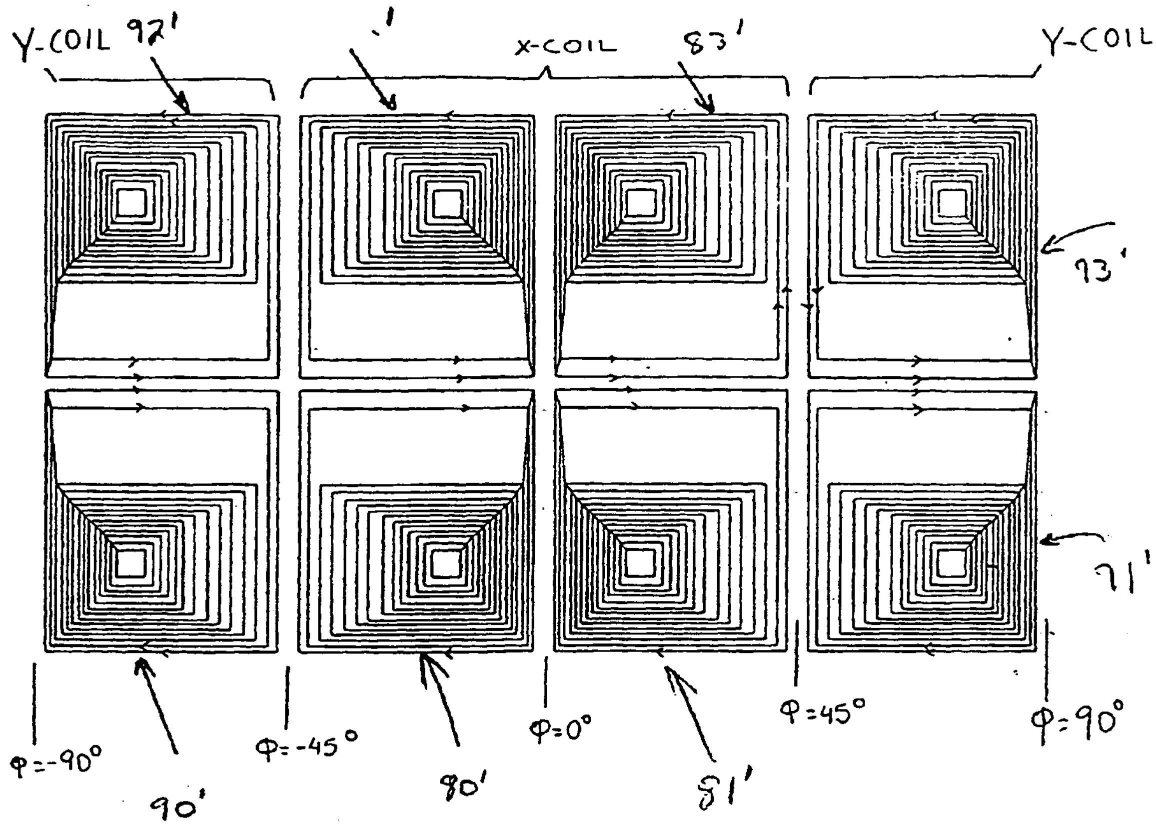


FIGURE 7

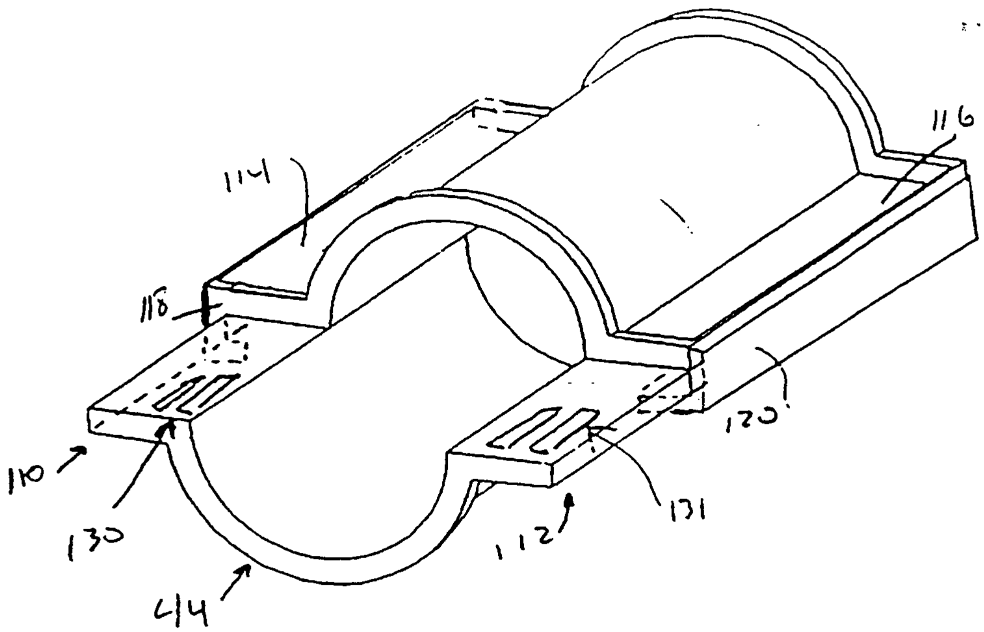


FIG 8 SLIDING TOP

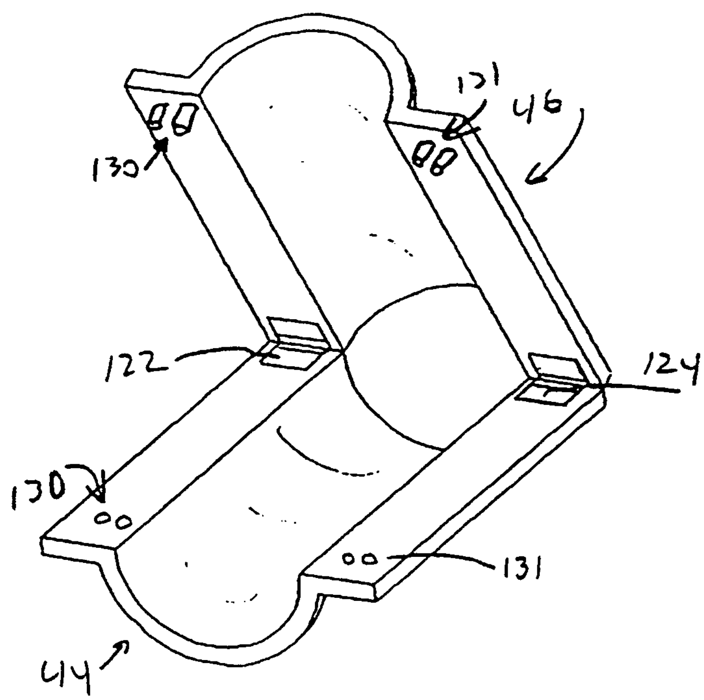


FIG. 9 HINGE DESIGN

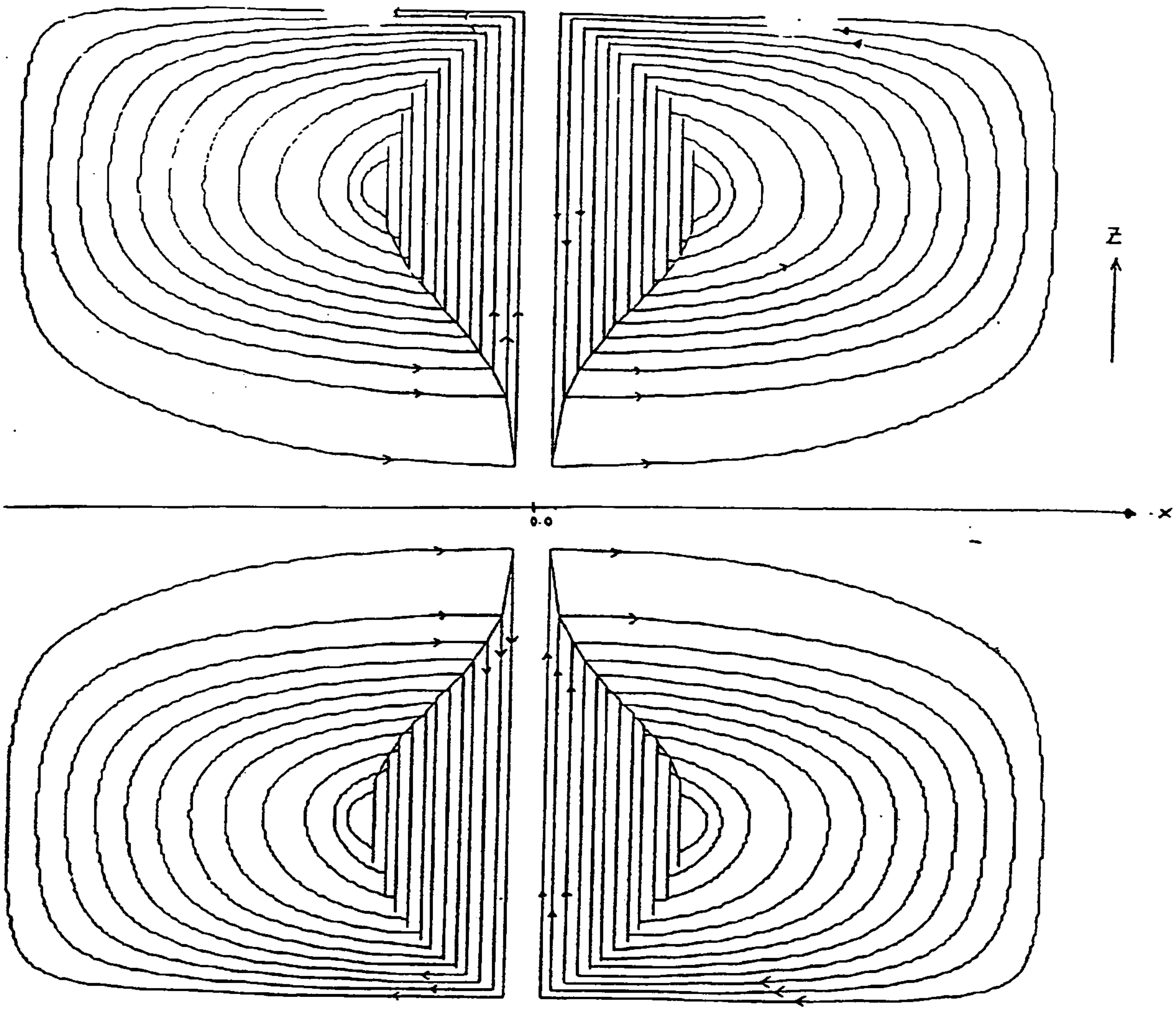
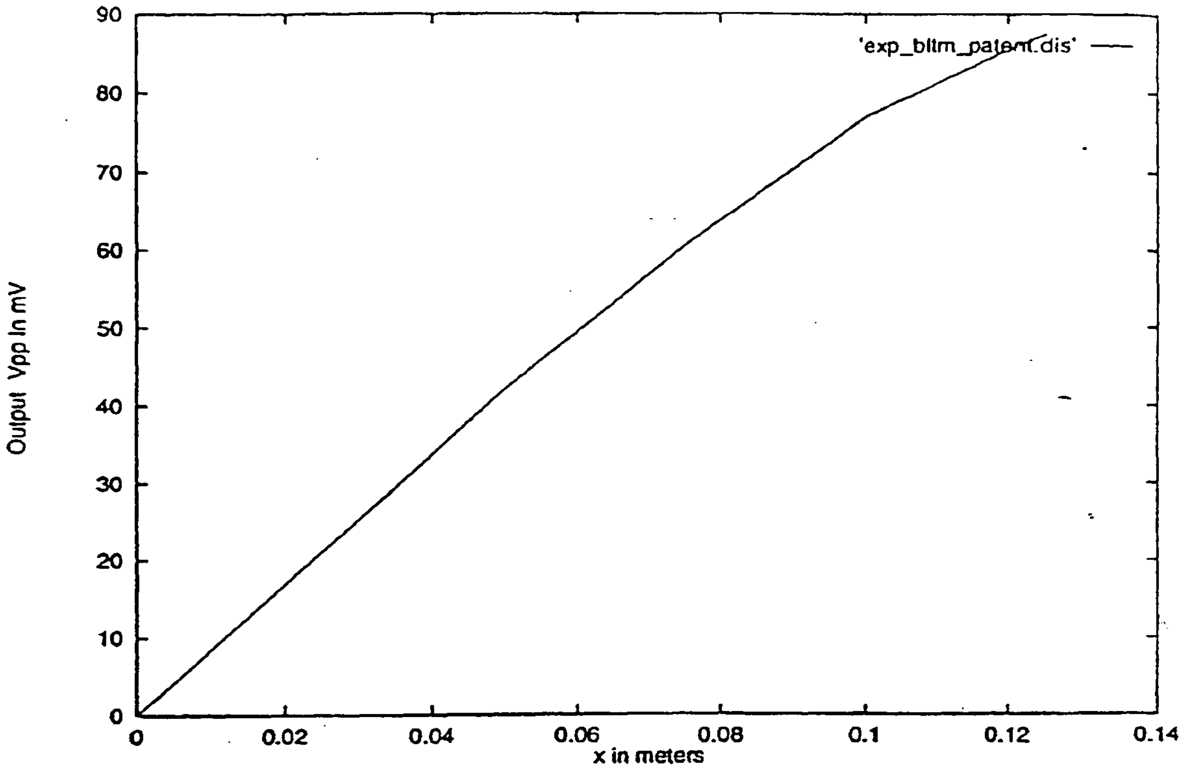


Illustration of the one-half of the discrete current distribution for Octapole X gradient coil. There are 4 quadrants per half coil and each quadrant has 12 loops with a common current of 199.53 A per loop.

FIG. 10



On-Axis behavior of the Octapole X gradient coil with a 3.5 cm Azimuthal Gap, based on the experimental measurements. The calculated gradient strength is 24.2 mT/m at 199.53 Amps

FIG. 11

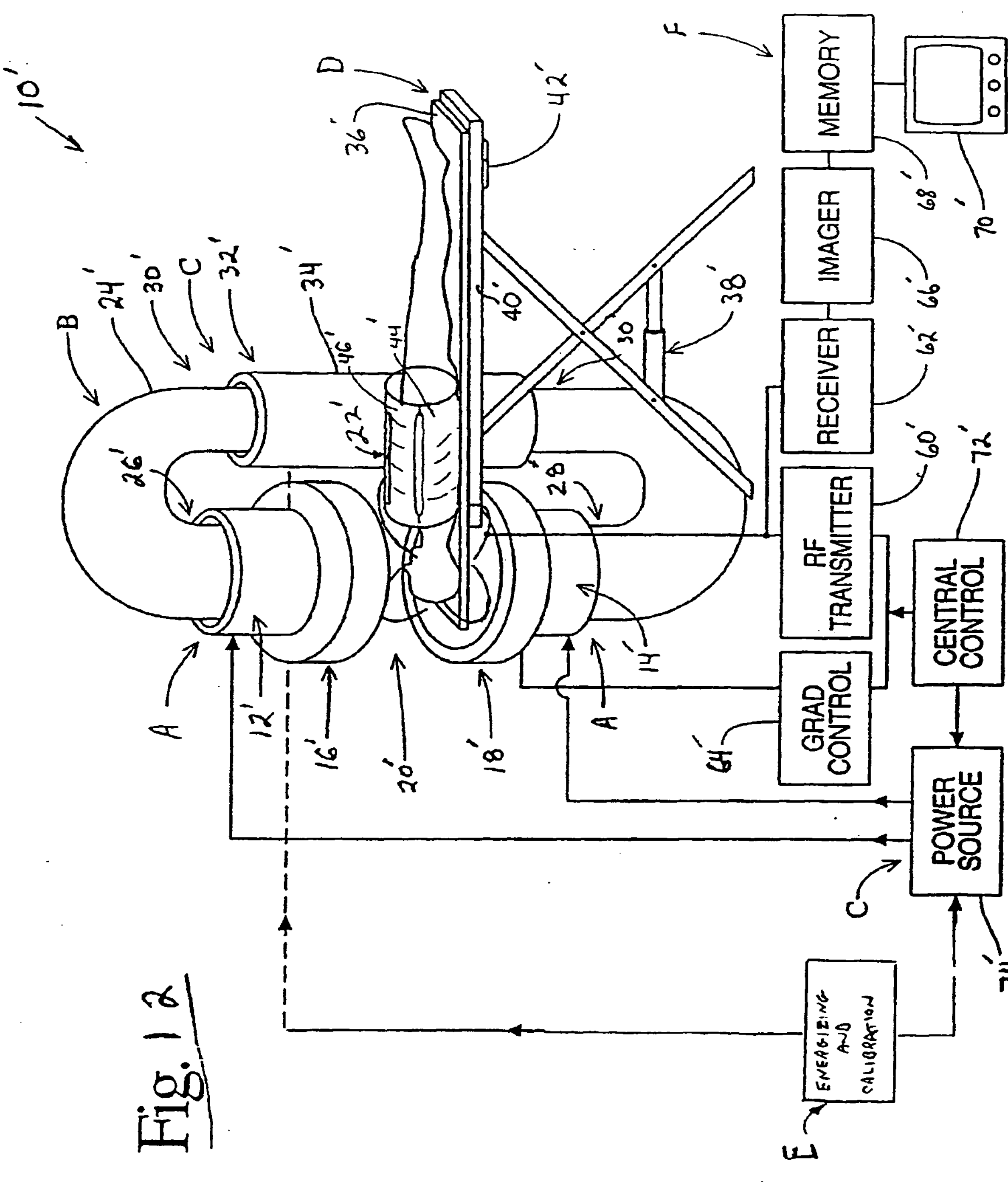

MEMORY
IMAGER
RECEIVER
RF TRANSMITTER
GRAD CONTROL
CENTRAL CONTROL
POWER SOURCE
ENERGIZING AND CALIBRATION

Fig. 12

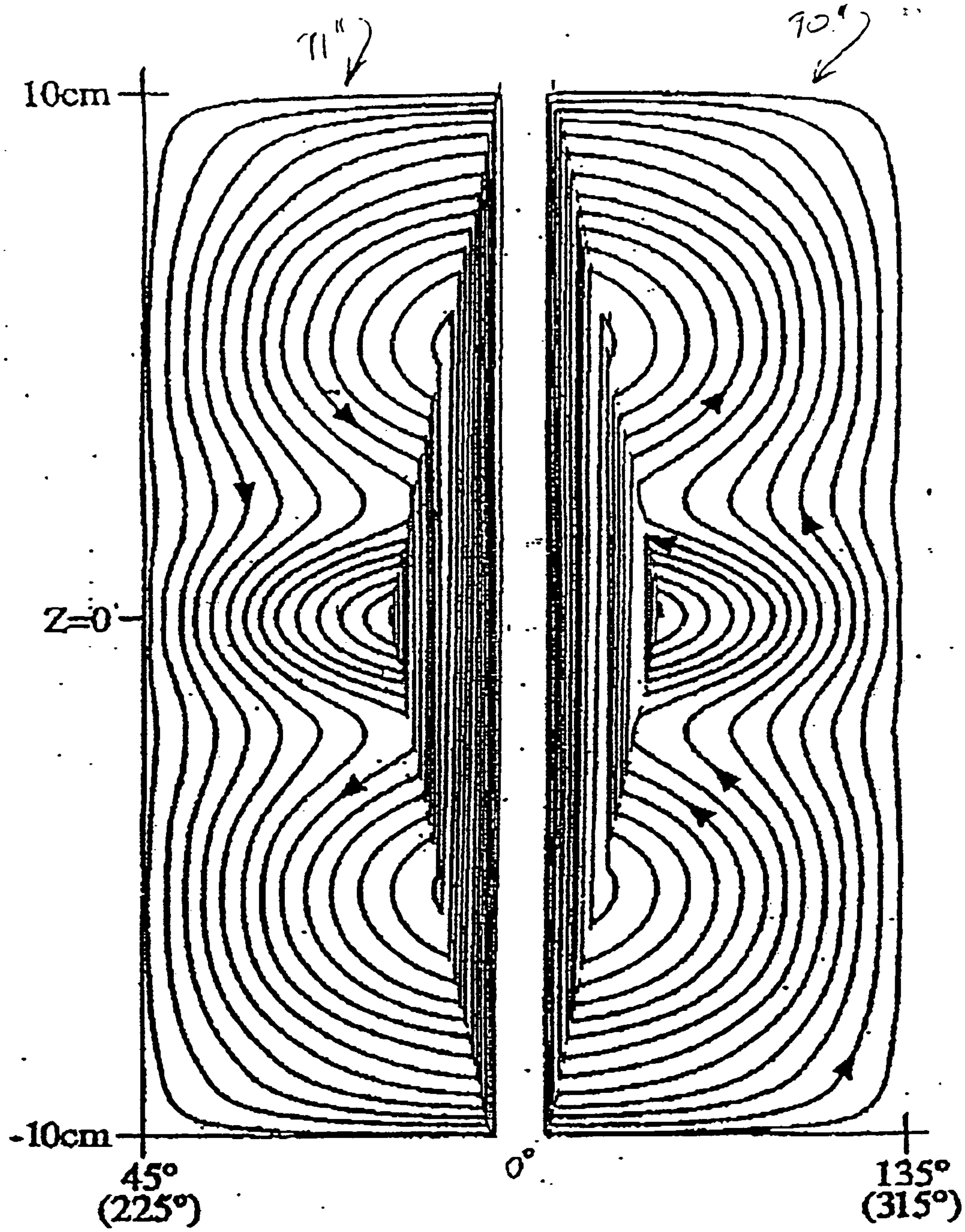


FIG. 13

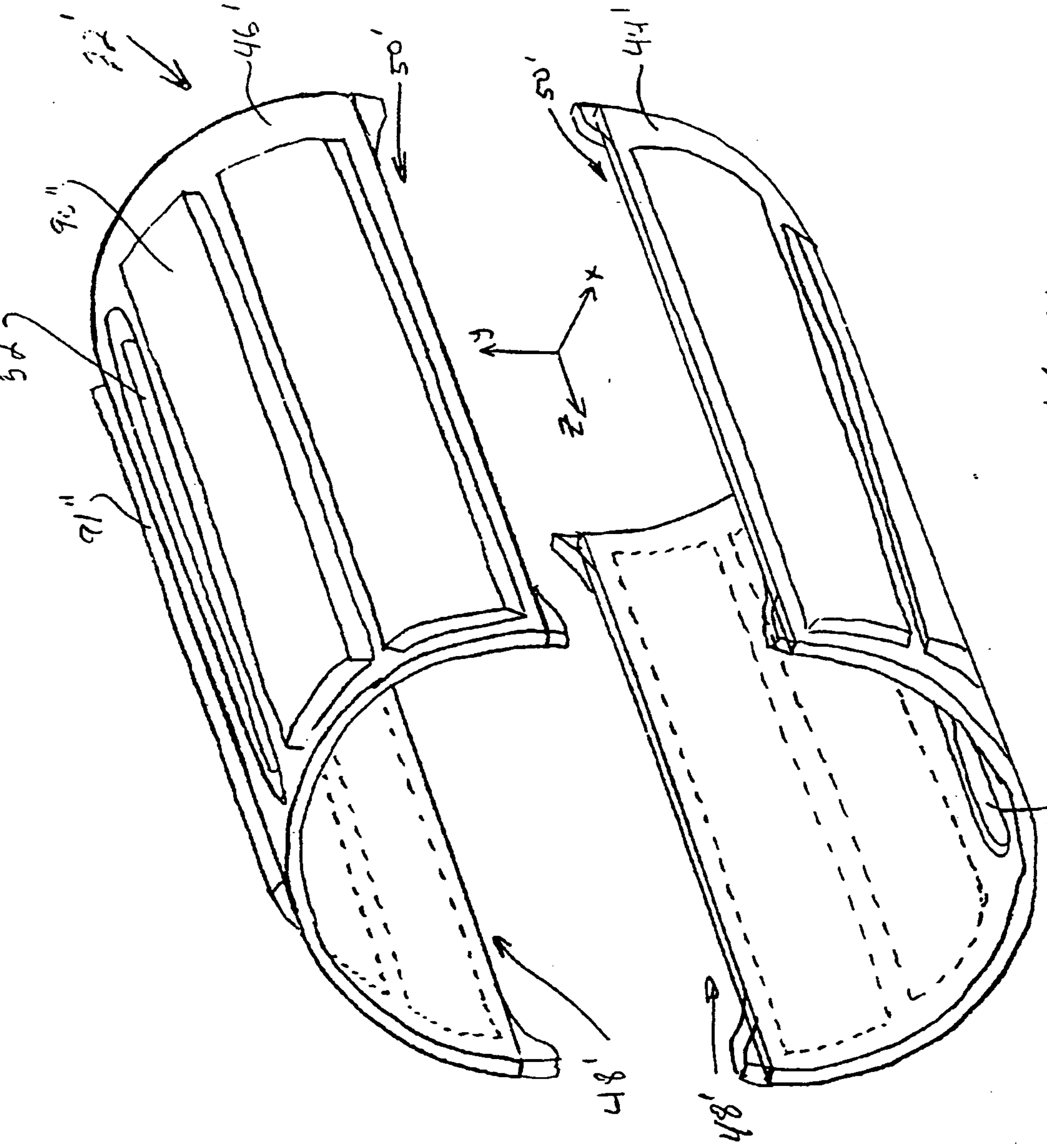


FIG. 14

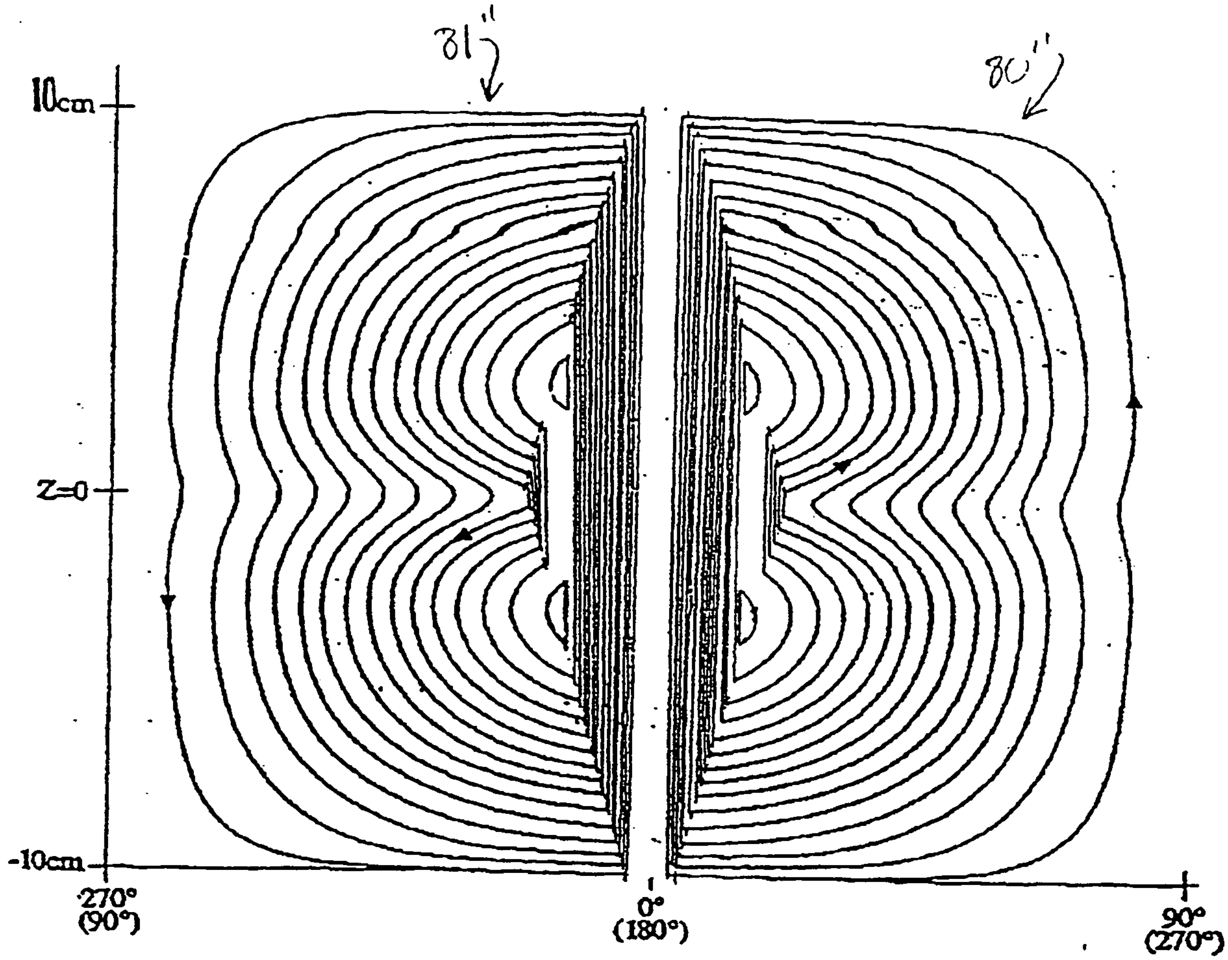


FIG. 15

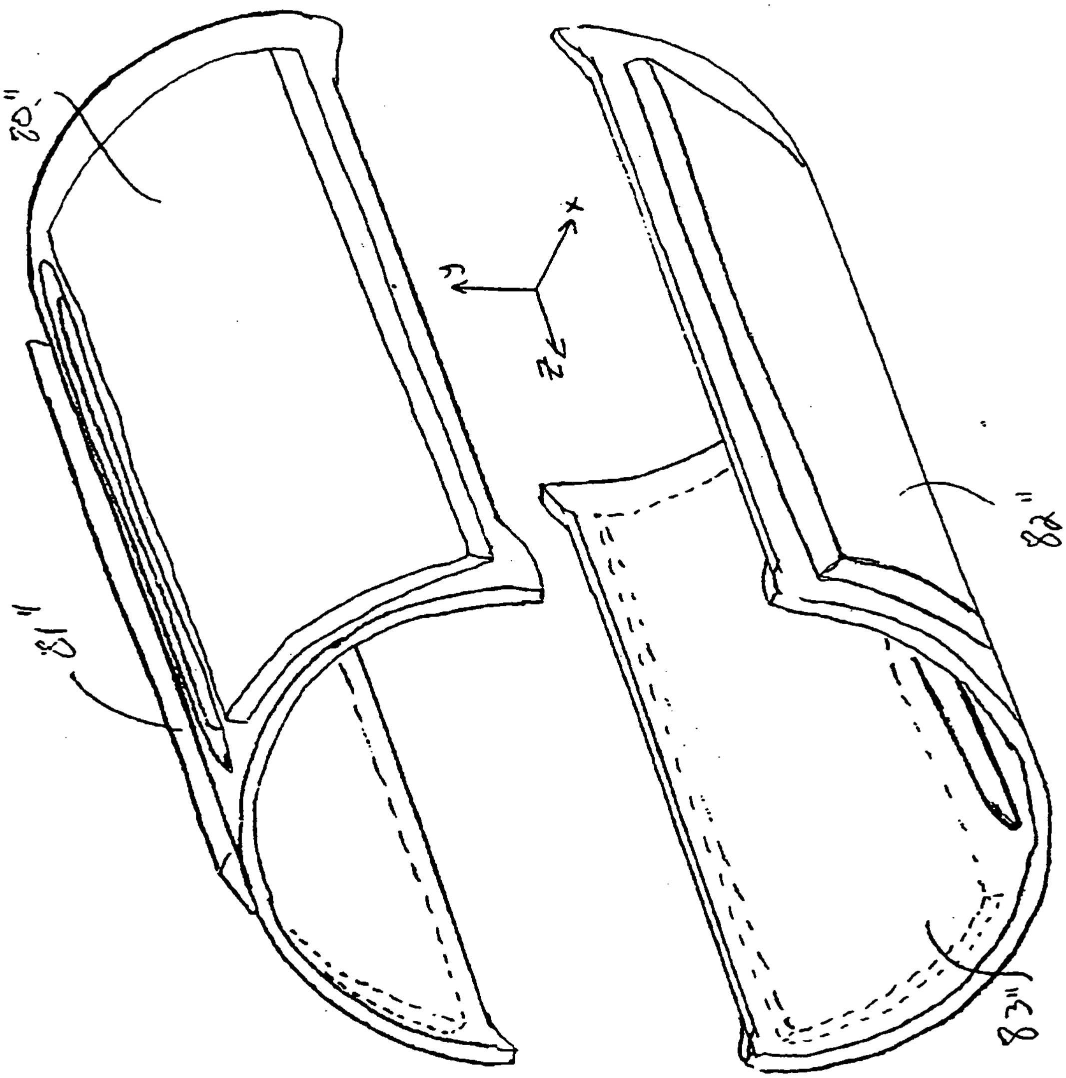


FIG. 16